A basic investigation of noise reduction for MEG data using independent component analysis based on correlation function

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1 Introduction

In the study of MEG, experimental tasks are restricted by the averaging procedure. For example, the analysis of the olfactory response is difficult, because it is not possible to perform averaging of many epochs without avoiding adaptation of the subjects to the stimuli. Moreover, it is reported that the rhythmic noise components remain after the averaging procedure and that it affects the signal source estimation [1,2]. Therefore, it is important to reduce the noise components. The principal component elimination method (PCEM) [3] and the signal-space projection (SSP) [4] have been used for noise reduction. It is concerned that the methods may reduce the signal, when the signal component and noise component is not orthogonal. Therefore, we applied the independent component analysis (ICA).

In this study, for the purpose of reducing the number of averaging epochs, we investigate a noise reduction using the ICA.

2 Method

The magnetic field measured by the multi-channel SQUID system is expressed by a vector **b**(t).

$$\mathbf{b}(t) = [b_1(t), b_2(t), \ldots, b_n(t)]^T \qquad (1)$$

Source signals are denoted by a vector **s**(t).

$$\mathbf{s}(t) = [s_1(t), s_2(t), \ldots, s_n(t)]^T \qquad (2)$$

The magnetic fields **b**(t) are represented as

$$\mathbf{b}(t) = \mathbf{A}\mathbf{s}(t) \qquad (3)$$

where **A** is the mixing matrix.

Here, **W** is the matrix which restores the independent sources **s**(t) from measured signals **b**(t).

$$\mathbf{y}(t) = \mathbf{W}\mathbf{b}(t) \qquad (4)$$

Therefore, the problem is to estimate **W** by the constraint with statistically independent **y**(t). We used the ICA algorithm to minimize cross-correlations [5,6] for estimating a constant matrix **W**.

2.1 Sphering

The eigenvalues $\mathbf{\Lambda}$ and eigenvectors **V** of the measurement data are denoted by

$$E\{\mathbf{b}(t)\mathbf{b}(t)^T\} = \mathbf{V}\mathbf{\Lambda}\mathbf{V}^T \qquad (5)$$

where $E\{\cdot\}$ denotes expectaion. The matrix $\mathbf{\Lambda}$ and **V** are reduced m-dimension by cumulative proportion.

$$\mathbf{\Lambda} = \begin{bmatrix} \lambda_1 & & & 0 \\ & \lambda_2 & & \\ & & \ddots & \\ 0 & & & \lambda_m \end{bmatrix}, \mathbf{V} = [\mathbf{v}_1, \mathbf{v}_2, \ldots, \mathbf{v}_m] \qquad (6)$$

Sphering is expressed by

$$\hat{\mathbf{b}}(t) = \sqrt{\mathbf{\Lambda}^{-1}}\mathbf{V}^T\mathbf{b}(t) \qquad (7)$$

2.2 Contrast function

We use the contrast function $F(\mathbf{C})$ based on the correlation structure of the signals. The rotation matrix **C** is obtained by minimizing the function.

$$F(\mathbf{C}) = \sum_{k=1}^{\tau}\sum_{i\neq j}\left|\left(\mathbf{C}E\{\hat{\mathbf{b}}(t)\hat{\mathbf{b}}(t+\tau_k)^T\}\mathbf{C}^T\right)_{ij}\right|^2 \qquad (8)$$

The JADE algorithm [7,8] is applied for the minimization of the contrast function. A constant matrix $\hat{\mathbf{W}}$ is obtained by $\hat{\mathbf{W}} = \mathbf{C}\sqrt{\mathbf{\Lambda}^{-1}}\mathbf{V}^T$ with estimated rotation matrix **C**. We can obtained estimated independent component $\hat{\mathbf{y}}(t)$ by equation (4) with $\hat{\mathbf{W}}$.

2.3 The selection of the noise components

The independent components are to be distinguished the magnetic fields which caused by the brain activity from the other magnetic fields. The criteria to distinguish them is as follows. Let us define the lead-field matrix as **L**, its signal subspace of the eigenvectors as follows

$$\mathbf{S}_\mathrm{s} = [\mathbf{s}_{\mathrm{s}1}, \mathbf{s}_{\mathrm{s}2}, \ldots, \mathbf{s}_{\mathrm{s}n}] \qquad (9)$$

where n is the number of 99.9% cumulative proportion [9]. The matrix $\hat{\mathbf{A}} = [\hat{\mathbf{a}}_1, \hat{\mathbf{a}}_2, \ldots, \hat{\mathbf{a}}_m]$ is pseudo inverse of $\hat{\mathbf{W}}$. The component $\hat{y}(t)$ is selected as noise when corresponding $\hat{\mathbf{a}}_m$ satisfies the following inequality

$$\left\|\frac{\hat{\mathbf{a}}_m^T}{\|\hat{\mathbf{a}}_m\|}\mathbf{S}_\mathrm{s}\right\| < 0.7 \qquad (10)$$

Such noise components $\hat{\mathbf{y}}(t)$ are reconstructed to magnetic fields. Then, the noise magnetic fields are removed from measured data.

3 Measurement

The non-averaging auditory evoked magnetic fields are analyzed by the above method. Four normal subjects (22-30 years male) participated in this study. The magnetic fields are measured in a magnetically shielded room using a 195-channel whole-head SQUID system for detecting vector component [10], and filtered on-line at 0.1-100 Hz and notch filter 50 Hz.

The sampling frequency and the number of the data acquisitions are 400Hz and 20000 points, respectively. After the data acquisition, non-averaged data were digitally filtered using a low-pass filter of 45Hz.

We used 1000Hz tone as auditory stimuli. The stimuli are delivered to right ear through a tube at an interval of 1.0sec.

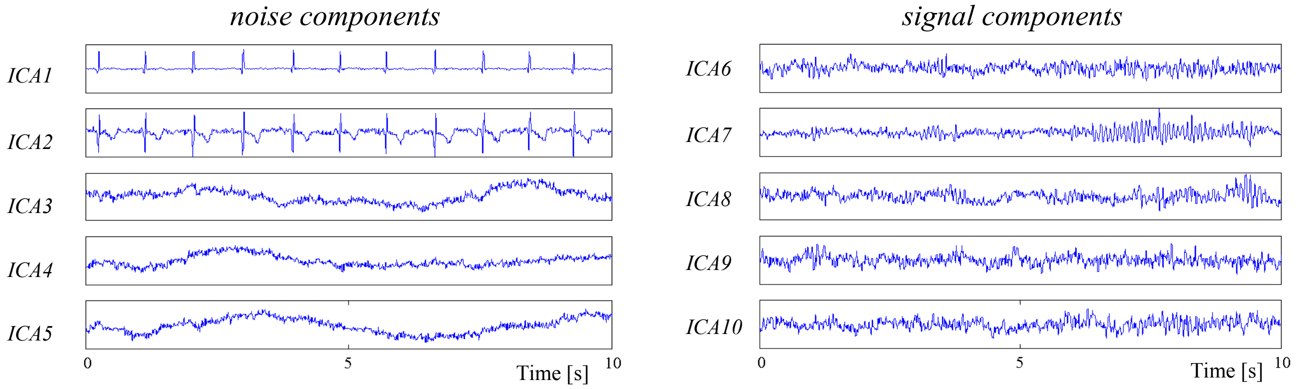
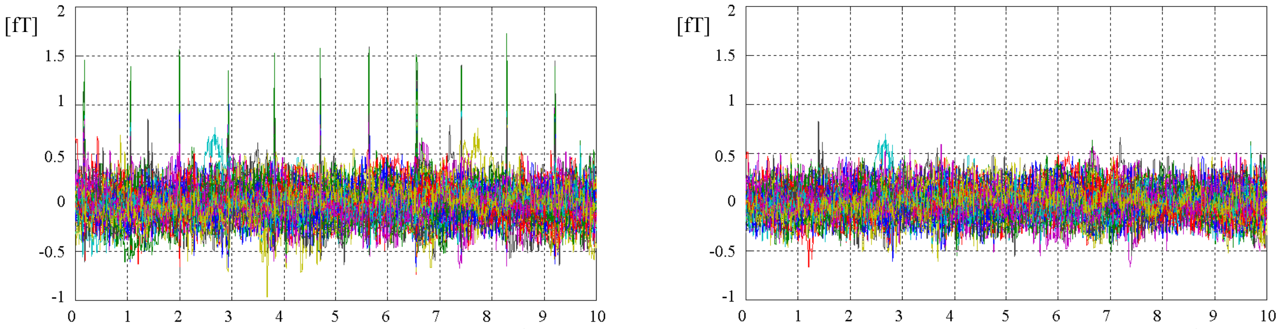
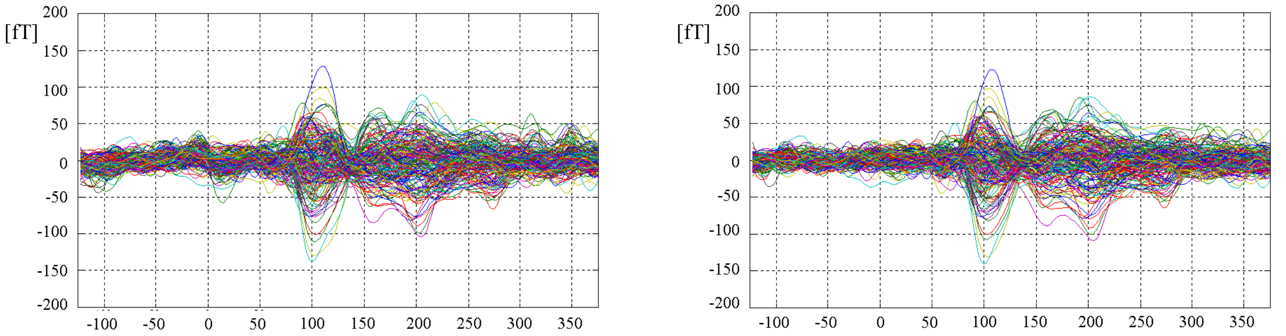


Figure 1: *The waveform of 10 components within extracted independent components. Noise components caused by cardiac activity are extracted (ICA1,ICA2).*



(a) *(b)*

Figure 2: (*a) The raw data of auditory magnetic fields. The magnetic fields which caused by cardiac activity is observed. (b) The magnetic fields obtained after noise reduction. A noise component caused by cardiac activity is removed by our method. The waveforms at 195 coil locations are superimposed.*



(a) *(b)*

Figure 3: (*a) Auditory evoked magnetic fields obtained by averaging 150 epochs. (b) Auditory evoked magnetic fields obtained by averaging 150 epochs after noise reduction.*

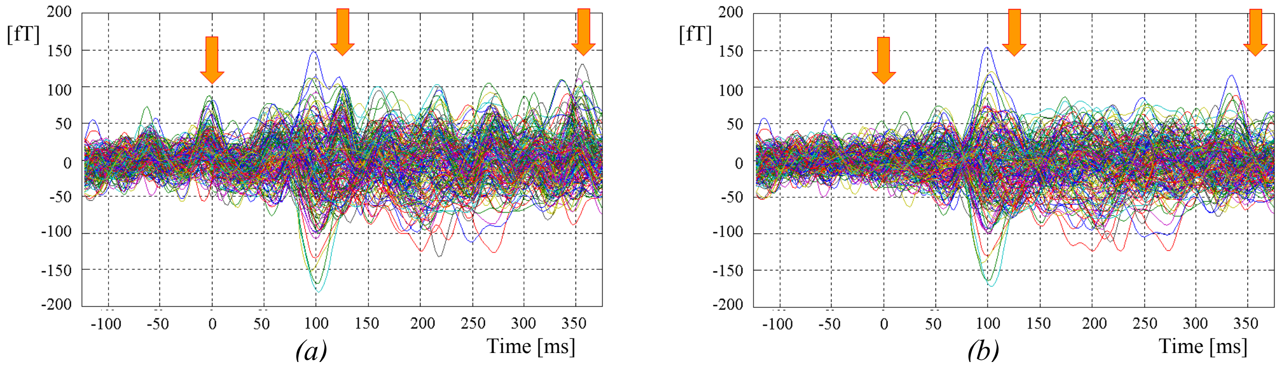


Figure 4: (*a) Auditory evoked magnetic fields obtained by averaging 30 epochs. (b) Auditory evoked magnetic fields obtained by averaging 30 epochs after noise reduction.*

4 Results

Independent components are extracted from the MEG data by using the ICA based on correlation function. Noise components caused by cardiac activity are extracted by the ICA (e.g., *ICA1* and *ICA2* in Figure 1). Besides the cardiac component, noise component seems to be caused by spontaneous brain activity is also extracted (e.g., *ICA7* in Figure 1). It was separated to signal components and noise components by applying our method (Figure 1).

The raw data and noise reduction data which superimposed the 195 waveforms is shown in Figure 2(a) and (b), respectively. Figure 2(b) shows that the noise component caused by cardiac activity is removed.

Figure 3 shows the evoked responses obtained by averaging 150 epochs (a) without the noise reduction and (b) with the noise reduction, respectively. Figure 3(b) shows that the waveforms of the signals are preserved with the noise reduction. Table 1 shows the location distance of the estimated dipole sources of evoked responses averaged 150 epochs with and without the noise reduction in 4 subjects. The location distance is about 1.0 mm in all subjects.

Table 1: *The location distance of the dipole estimated from averaging 150 epochs with and without the noise reduction in 4 subjects.*

Subjects	S1	S2	S3	S4
Location distance [mm]	1.0	1.3	1.3	1.3

Figure 4(a) shows the evoked responses obtained by averaging 30 epochs. Noise component caused by cardiac activity, which is disappeared in the result of averaging 150 epochs, is remained in that of 30 epochs. However, it is removed by using our method with 30 epochs (Figure 4(b)).

Evoked responses of 150 epochs obtained from a subject are divided into 5 sections of 30 epochs. Figure 5 shows the location distances between the dipoles estimated from 150 epochs and from averaging 30 epochs in each section. In the figure the filled and open bars denote without and with the noise reduction, respectively. Figure 5 shows that the accuracy of the signal source localization is improved.

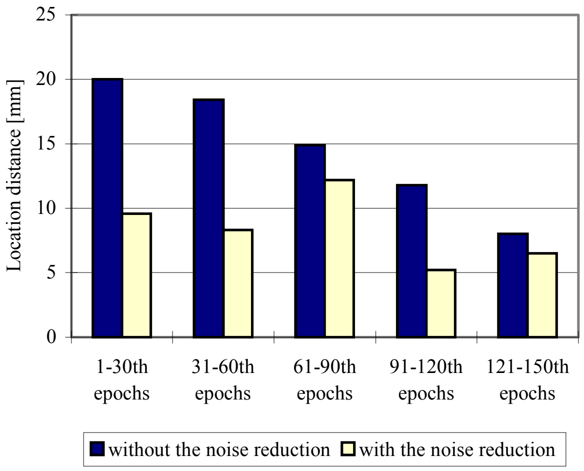


Figure 5: *The location distances between the dipoles estimated from 150 epochs and from 30 epochs in each section.*

Figure 6 shows the average location distances of 150 epochs and 30 epochs in each subject. The location distances are improved in 3 subjects except for subject S4.

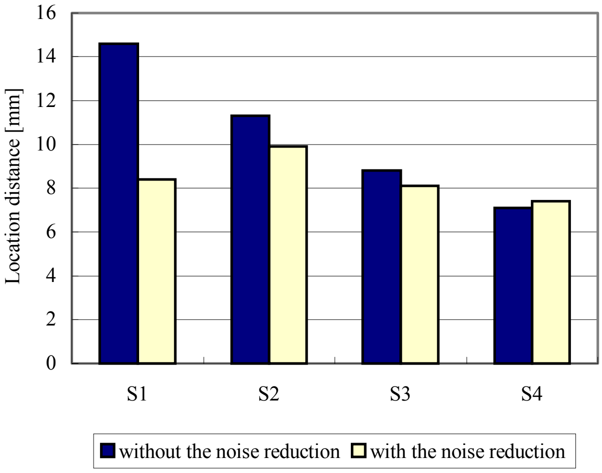


Figure 6: *The average location distances of 150 epochs and 30 epochs in each subject.*

5 Discussion

In previous studies, the method for choosing the noise components from extracted independent components are not established. On the other hand, it is possible to separate the signal components and the noise components automatically by our method. It is confirmed that the noise reduction by the ICA does not affect to the result of the localization (Table 1). In addition, the result of signal source localization of the averaged 30 epochs shows that the accuracy of the signal source estimation is improved. Our method is effective for the noise reduction of MEG data. However, it is not possible to recognize noise components arise from the brain activity which are not related the task. It is necessary to find the method for distinguishing those components from the signal components.

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References

1. K. Sekihara, Y. Ogura, and M. Hotta, “Maximum-likelihood estimation of current-dipole parameters for data obtained using multichannel magnetometer”, *IEEE Trans. Biomed. Eng.*, **39**, 558-562, 1992.
2. K. Sekihara, F. Takeuchi, S. Kuriki, and H. Koizumi, “Reduction of brain noise influence in evoked neuromagnetic source localization using noise spatial correlation”, *Phys. Med. Biol.*, **39**, 937-946, 1994.
3. T. Kobayashi and S. Kuriki, “Principal component elimination method for the improvement of S/N in evoked neuromagnetic field measurements”, *IEEE trans. Biomed. Eng.*, **46**, 951-958, 1999.
4. M.A.Uusitalo and R.J.Ilmoniemi, “The signal-space projection method for separating MEF or EEG into components”, *Med. & Biol. Eng.* & Comput., **35**, 135-140, 1997.
5. L. Molgedey and H.G. Schuster, “Separation of a mixture of independent signals using time delayed correlations”, *Phys. Rev. Lett.*, **72**, 3634-3637, 1994.
6. S. Ikeda and N. Murata, ”A method of blind separation based on temporal structure of signals”, In *Proceedings of The Fifth International Conference on Neural Information Processing (ICONIP'98 Kitakyushu)*, pp. 737-742, 1998.
7. J.F. Cardoso and A. Souloumiac, “Blind beamforming for non gaussian signals”, *IEE Proceedings F*, **140**, 362-370, 1993.
8. J.F. Cardoso and A. Souloumiac, “Jacobi angles for simultaneous diagonalization”, *SIAM J. Mat. Anal. Appl.*, **17**, 161-164, 1996.
9. M. Kawakatsu, K. Kobayashi, Y. Uchikawa, M. Adachi, M. Saito and M. Kotani, “A noise reduction using signal-subspace method”, *this volume.*
10. M. Kawakatsu, K. Kobayashi, Y. Uchikawa, and M. Kotani, “A whole-head 195-channel SQUID MEG system for detecting vector component”, in *Recent Advances in Biomagnetism*, T. Yoshimoto, M. Kotani, S. Kuriki, H. Karibe, and N. Nakasato, Eds. Sendai: Tohoku University Press, 1999, pp. 31-34.